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(54) **MRI COMPATIBLE TACHYCARDIA LEAD**

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CPC ***A61N 1/0563*** (2013.01); ***A61N 1/05*** (2013.01); *A61N 1/375* (2013.01); *A61N 2001/0585* (2013.01); *A61N 2001/086* (2013.01)

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CPC A61N 1/056; A61N 1/05; A61N 1/3752; A61N 1/0563; A61N 2001/086

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See application file for complete search history.

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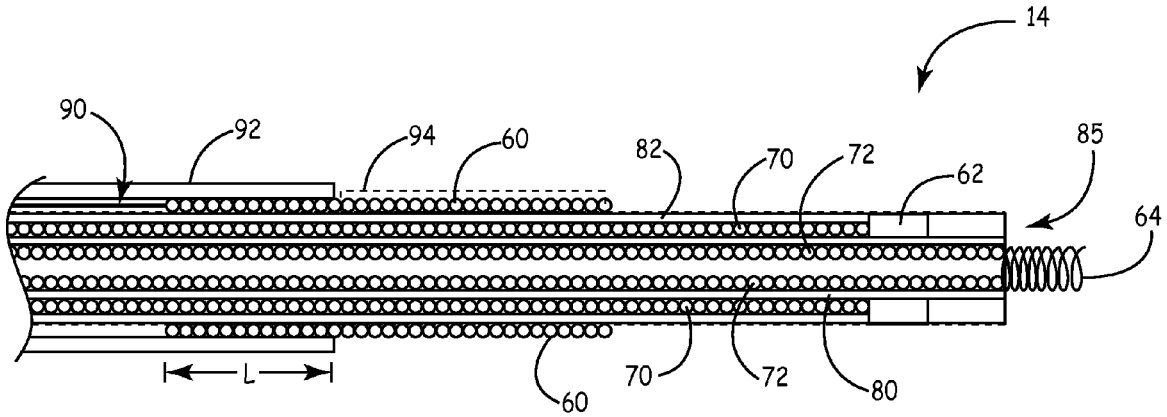
(57) **ABSTRACT**

A medical device lead includes a proximal connector configured to couple the lead to a pulse generator, an insulative lead body extending distally from the proximal connector, and a conductor assembly extending distally from the proximal connector within the lead body. The conductor assembly includes a conductor having a proximal end electrically coupled to the connector and a distal end electrically coupled to a defibrillation coil. A first portion of the defibrillation coil is exposed at an outer surface of the medical device lead and a second portion of the defibrillation coil is insulated at the outer surface of the medical device lead.

6 Claims, 4 Drawing Sheets



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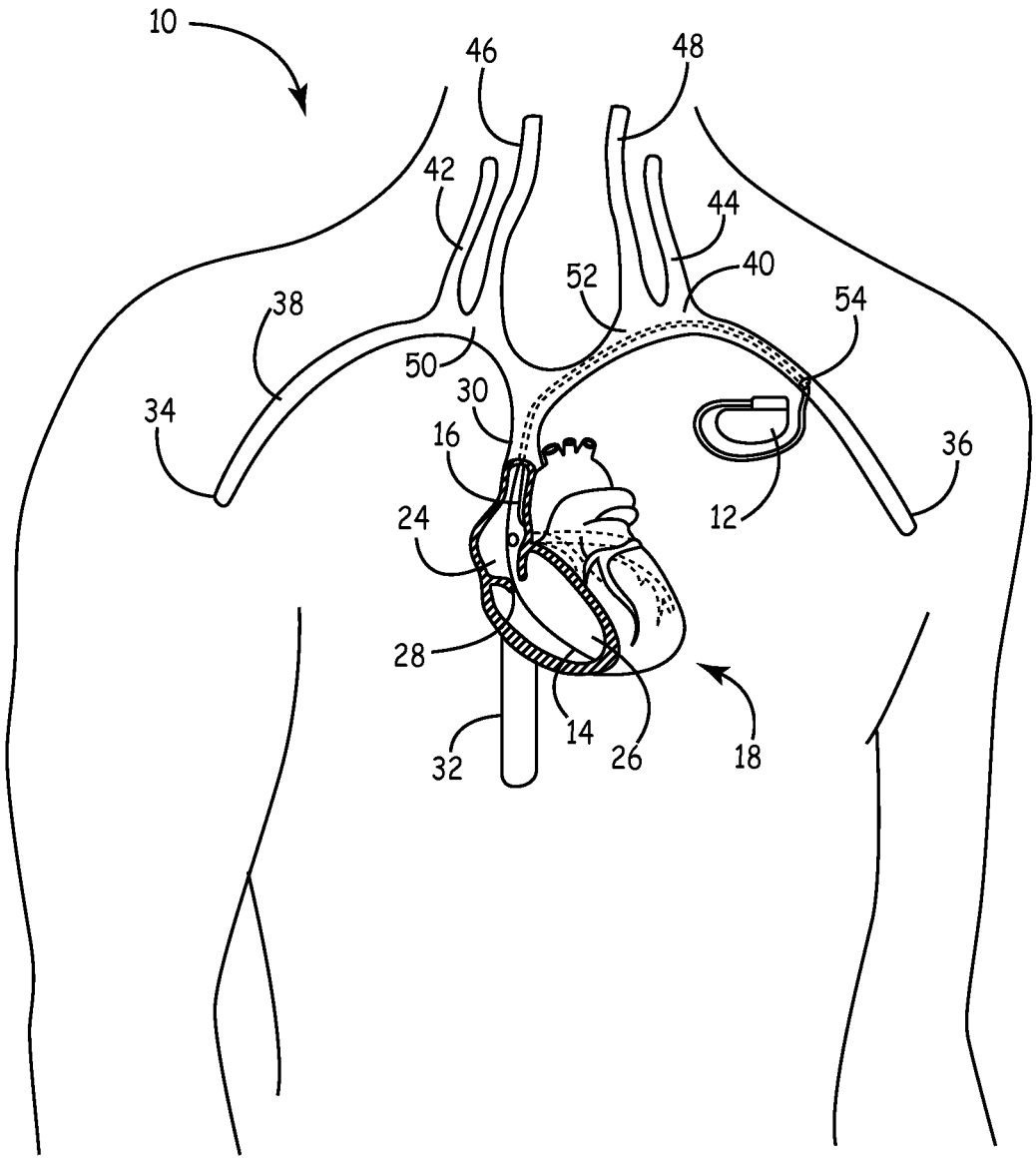

FIG. 1

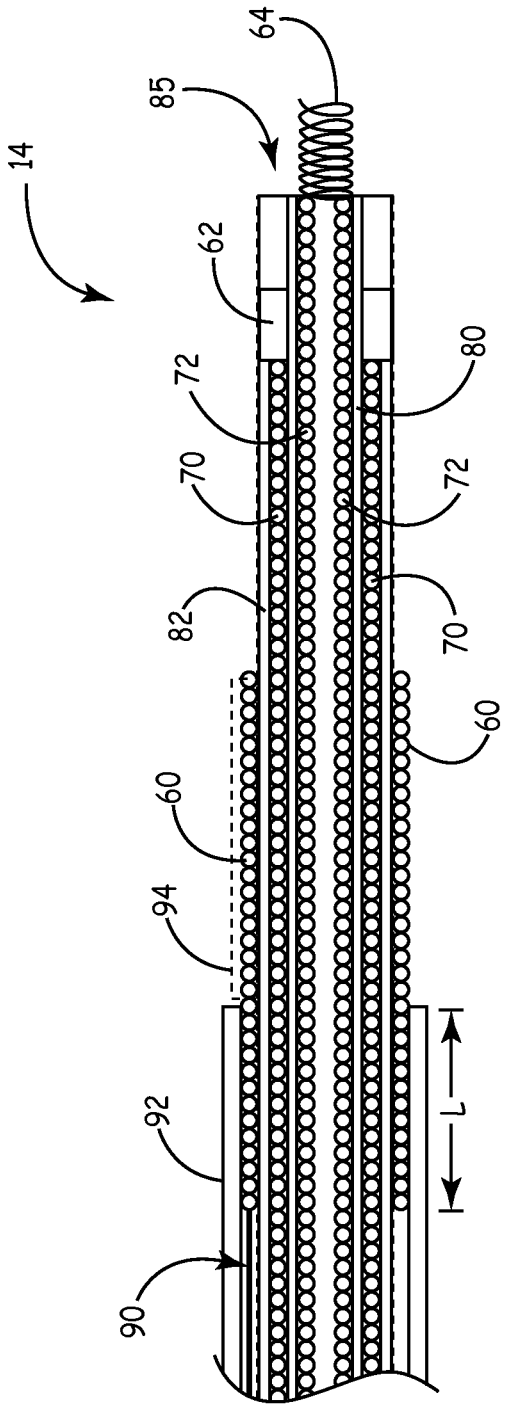

FIG. 2

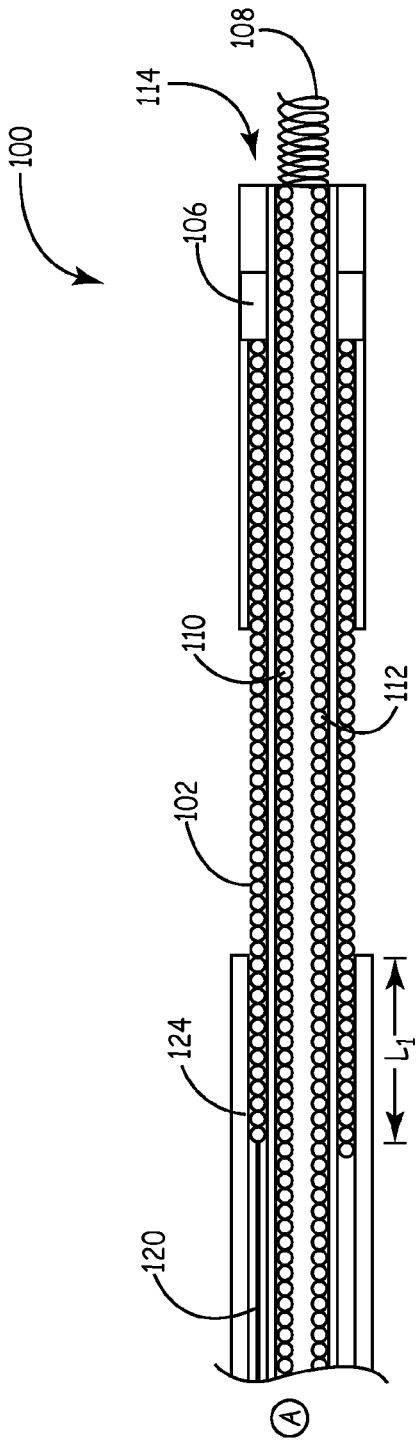
100
114
108
106
110
112
102
124
120
L_1
A

FIG. 3A

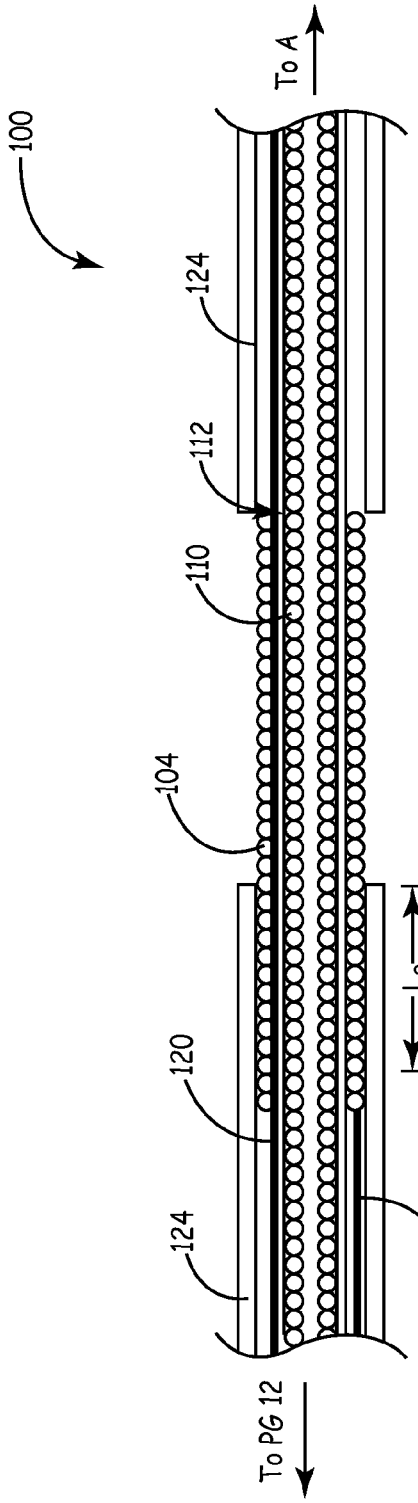
100
To A
124
112
110
104
120
L_2
124
122
To PG 12

FIG. 3B

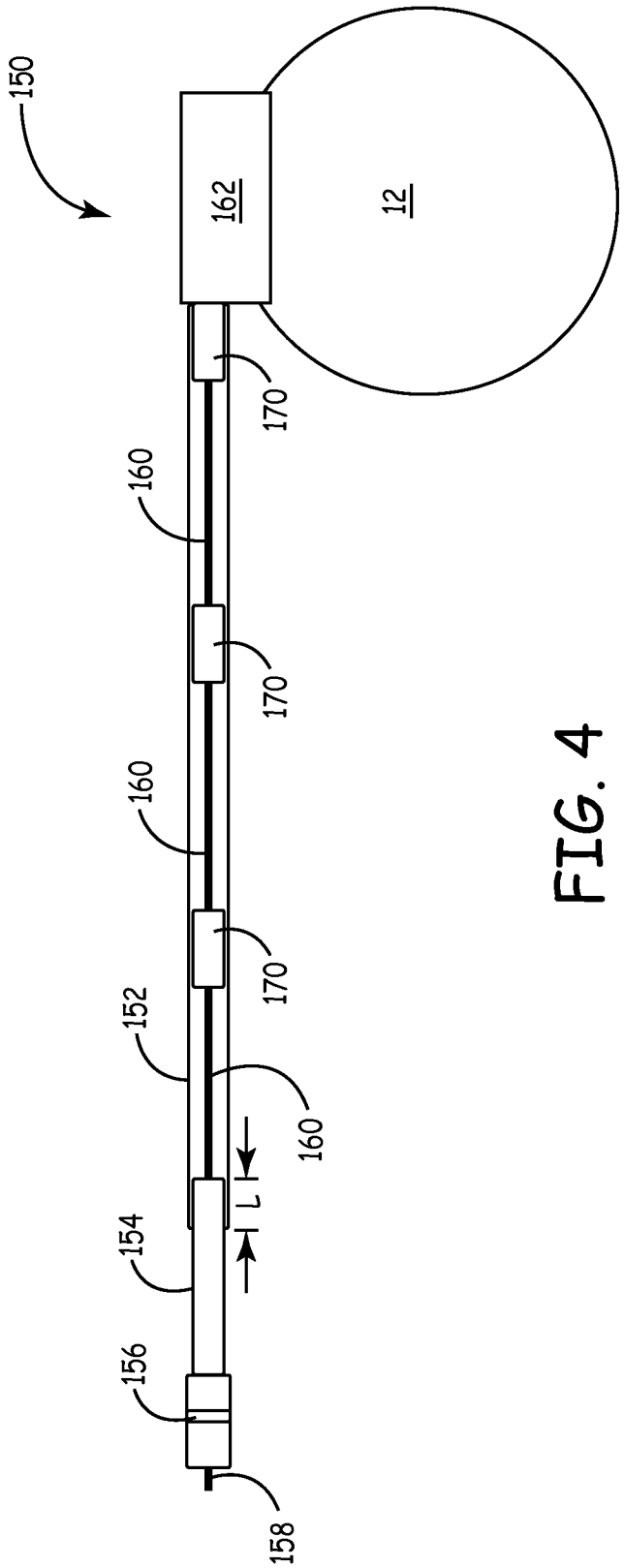

FIG. 4

MRI COMPATIBLE TACHYCARDIA LEAD

CROSS-REFERENCE TO RELATED APPLICATION

This application claims priority to Provisional Application No. 61/252,915, filed Oct. 19, 2009, which is herein incorporated by reference in its entirety.

TECHNICAL FIELD

The present invention relates to implantable medical devices. More particularly, the present invention relates to MRI-compatible tachycardia lead constructions.

BACKGROUND

Magnetic resonance imaging (MRI) is a non-invasive imaging procedure that utilizes nuclear magnetic resonance techniques to render images within a patient's body. Typically, MRI systems employ the use of a magnetic coil having a magnetic field strength of between about 0.2 to 3 Teslas. During the procedure, the body tissue is briefly exposed to RF pulses of electromagnetic energy in a plane perpendicular to the magnetic field. The resultant electromagnetic energy from these pulses can be used to image the body tissue by measuring the relaxation properties of the excited atomic nuclei in the tissue.

During imaging, the electromagnetic radiation produced by the MRI system may be picked up by implantable device leads used in implantable medical devices such as pacemakers or cardiac defibrillators. This energy may be transferred through the lead to the electrode in contact with the tissue, which may lead to elevated temperatures at the point of contact. The degree of tissue heating is typically related to factors such as the length of the lead, the conductivity or impedance of the lead, and the surface area of the lead electrodes. Exposure to a magnetic field may also induce an undesired voltage on the lead.

SUMMARY

The present invention relates to a medical device lead including a proximal connector configured to couple the lead to a pulse generator, an insulative lead body extending distally from the proximal connector, a conductor assembly extending distally from the proximal connector within the lead body. The conductor assembly includes a conductor having a proximal end electrically coupled to the connector and a distal end electrically coupled to a defibrillation coil. A first portion of the defibrillation coil is exposed at an outer surface of the medical device lead and a second portion of the defibrillation coil is insulated at the outer surface of the medical device lead.

In another aspect, a medical device lead includes a first connector configured to couple the lead to a pulse generator, an insulative lead body extending distally from the first connector, and a first conductor extending distally from the first connector within the lead body and having a proximal end electrically coupled to the first connector. A first defibrillation coil is exposed at an outer surface of the medical device lead, and a first high impedance coil is connected between the first conductor and the first defibrillation coil. The first high impedance coil is insulated at the outer surface of the medical device lead and has an impedance greater than the first conductor.

In a further aspect, a medical device lead includes one or more proximal connectors each configured to couple to a pulse generator, an insulative lead body extending distally from the one or more proximal connectors, and one or more conductors each extending distally from and electrically connected to one of the one or more proximal connectors. One or more defibrillation coils are each connected to a distal end of one of the one or more conductors. A first portion of each defibrillation coil is exposed at an outer surface of the medical device lead and a second portion of each defibrillation coil is insulated at the outer surface of the medical device lead.

While multiple embodiments are disclosed, still other embodiments of the present invention will become apparent to those skilled in the art from the following detailed description, which shows and describes illustrative embodiments of the invention. Accordingly, the drawings and detailed description are to be regarded as illustrative in nature and not restrictive.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. **1** is a schematic view of a cardiac rhythm management (CRM) system including a pulse generator and a lead implanted in a patient's heart according to an embodiment of the present invention.

FIG. **2** is a cross-sectional view of a distal portion of a lead according to an embodiment of the present invention including a defibrillation coil that is partially insulated.

FIG. **3**A is a cross-sectional view of a distal portion of a lead according to another embodiment of the present invention including two defibrillation coils that are each partially insulated.

FIG. **3**B is a cross-sectional view of a portion of the lead proximal to the distal portion of the lead shown in FIG. **3**A.

FIG. **4** is a schematic view of an implantable medical device including a lead having a plurality of high impedance coils in series with a conductive cable that extends through the lead body according to a further embodiment of the present invention.

While the invention is amenable to various modifications and alternative forms, specific embodiments have been shown by way of example in the drawings and are described in detail below. The intention, however, is not to limit the invention to the particular embodiments described. On the contrary, the invention is intended to cover all modifications, equivalents, and alternatives falling within the scope of the invention as defined by the appended claims.

DETAILED DESCRIPTION

FIG. **1** is a schematic view of a cardiac rhythm management (CRM) system **10** according to an embodiment of the present invention. As shown in FIG. **1**, the CRM system **10** includes a pulse generator **12** coupled to a plurality of leads **14**, **16** deployed in a patient's heart **18**. As further shown in FIG. **1**, the heart **18** includes a right atrium **24** and a right ventricle **26** separated by a tricuspid valve **28**. During normal operation of the heart **18**, deoxygenated blood is fed into the right atrium **24** through the superior vena cava **30** and the inferior vena cava **32**. The major veins supplying blood to the superior vena cava **30** include the right and left axillary veins **34** and **36**, which flow into the right and left subclavian veins **38** and **40**. The right and left external jugular **42** and **44**, along with the right and left internal jugular **46** and **48**, join the right and left subclavian veins **38** and **40** to form the right and left brachiocephalic veins **50** and **52**, which in turn combine to flow into the superior vena cava **30**.

The leads **14**, **16** operate to convey electrical signals and stimuli between the heart **18** and the pulse generator **12**. In the illustrated embodiment, the lead **14** is implanted in the right ventricle **26**, and the lead **16** is implanted in the right atrium **24**. In other embodiments, the CRM system **10** may include additional leads, e.g., a lead extending into a coronary vein for stimulating the left ventricle in a bi-ventricular pacing or cardiac resynchronization therapy system. As shown, the leads **14**, **16** enter the vascular system through a vascular entry site **54** formed in the wall of the left subclavian vein **40**, extend through the left brachiocephalic vein **52** and the superior vena cava **30**, and are implanted in the right ventricle **26** and right atrium **24**, respectively. In other embodiments of the present invention, the leads **14**, **16** may enter the vascular system through the right subclavian vein **38**, the left axillary vein **36**, the left external jugular **44**, the left internal jugular **48**, or the left brachiocephalic vein **52**.

The pulse generator **12** is typically implanted subcutaneously within an implantation location or pocket in the patient's chest or abdomen. The pulse generator **12** may be any implantable medical device known in the art or later developed, for delivering an electrical therapeutic stimulus to the patient. In various embodiments, the pulse generator **12** is a pacemaker, an implantable cardiac defibrillator, and/or includes both pacing and defibrillation capabilities. The portion of the leads **14**, **16** extending from the pulse generator **12** to the vascular entry site **54** are also located subcutaneously or submuscularly. The leads **14**, **16** are each connected to the pulse generator **12** via proximal connectors. Any excess lead length, i.e., length beyond that needed to reach from the pulse generator **12** location to the desired intracardiac implantation site, is generally coiled up in the subcutaneous pocket near the pulse generator **12**.

FIG. **2** is a cross-sectional view of the lead **14** according to an embodiment of the present invention. The lead **14** includes a defibrillation coil **60**, and pacing or sensing electrodes **62** and **64**. The defibrillation coil **60** may be used to deliver a high voltage therapy signal to a portion of the heart **18**. The pacing or sensing electrodes **62** and **64** may be used for pacing, sensing, or both. In the embodiment shown, the electrode **62** is a ring electrode, and the electrode **64** includes a fixation helix. In some embodiments, the electrodes **62** and **64** include platinum or titanium coated with a combination of iridium oxide (IrOx), titanium/nickel (Ti/Ni), black platinum (Pt black), or tantalum oxide (TaO). The defibrillation coil **60** and the pacing or sensing electrodes **62** and **64** are located near a distal end portion of the lead **14**. In alternative embodiments, the defibrillation and pacing or sensing electrodes are located elsewhere on the lead **14**. The lead **14** may also alternatively include fewer or more electrodes.

The electrode **62** is coupled to a first conductive coil **70**, and the electrode **64** is coupled to a second conductive coil **72**. The second conductive coil **72** is surrounded by an insulative layer **80** to insulate the conductive coil **72** from other elements of the lead **14**. In some embodiments, the insulative layer **80** extends from the proximal end to the distal end of the lead **14**. An insulative layer **82** is also formed around the first conductor **70**. In some embodiments, the insulative layer **82** extends from the proximal end of the lead **14** to the electrode **62**. With this arrangement, the electrode **62** is exposed at the outer surface of the lead **14** to allow contact with adjacent tissue. The insulative layers **80** and **82** may be comprised of, for example, silicone material, Teflon, expanded polytetrafluoroethylene (ePTFE), polytetrafluoroethylene (PTFE), ethylene tetrafluoroethylene (ETFE), or another suitable non-conductive material. The electrodes **62** and **64**, the conductive coils **70** and **72**, and the insulative layers **80** and **82** combine to form the low voltage pacing/sensing portion **85** of the lead **14**.

The first conductive coil **70** and the second conductive coil **72** extend through the lead **14** and are adapted for connection to the pulse generator **12** at the proximal end of the lead **14**. In some embodiments, the first conductive coil **70** and the second conductive coil **72** are each coupled to a proximal connector at the proximal end of the lead **14**. The connectors at the proximal end of the lead **14** are sized and shaped to interface with a connector block or other component of the pulse generator **12**. The signals carried by the first conductive coil **70** and the second conductive coil **72** may be independently controlled by the pulse generator **12** such that different signals may be delivered to and/or received from the electrodes **62** and **64**.

The defibrillation coil **60** is coupled to a conductive cable **90**, which extends through the lead **14** and is adapted for connection to the pulse generator **12** at the proximal end of the lead **14**. The conductive cable **90** may extend through the lead **14** in a lumen parallel to the conductive coils **70** and **72**. The conductive cable **90** is surrounded by an insulating layer **92** at an exterior surface of the lead **14**. In some embodiments, the conductive cable **90** is coupled to a proximal connector at the proximal end of the lead **14** that is sized and shaped to interface with a connector block or other component of the pulse generator **12**. The conductive cable **90** delivers a high voltage defibrillation signal from the pulse generator **12** to the defibrillation coil **60**. The lead **14** is arranged in the heart **18** such that the signal delivered by the defibrillation coil **60** depolarizes a critical mass of the heart muscle, terminates an arrhythmia, and allows normal sinus rhythm to be reestablished.

In a magnetic resonance imaging (MRI) environment, the radio frequency (RF) fields can induce a current in the conductive elements of the lead **14**. This current may then be dissipated at the point of contact between the lead electrodes and adjacent tissue, resulting in elevated temperatures in the tissue. To reduce the RF current that is transmitted to the defibrillation coil **60** by the conductive cable **90**, a length L of the defibrillation coil **60** is insulated at the exterior surface of the lead **14** by insulating layer **92**. The insulated length L of the defibrillation coil **60** acts as an RF filter between the conductive cable **90** and the exposed portion of the defibrillation coil **60**. More specifically, the inductance of a coil is directly proportional to the square of the radius of the coil. Thus, the inductance of the defibrillation coil **60** is large due to its large diameter. In some embodiments, the outside diameter of the defibrillation coil **60** is in the range of about 0.08 to 0.12 inch (about 2.0 to 3.0 mm). Consequently, the insulated length L of the defibrillation coil **60** reduces the amount of MRI-induced energy that is transmitted to the defibrillation coil **60** via the conductive cable **90**. In some embodiments, the proximal and distal ends of the exposed portion of the defibrillation coil **60** are short circuited with an optional low impedance connection **94** (shown in phantom) to evenly distribute the high voltage signal across the exposed portion.

The inductance of a coil is also directly proportional to the square of the number of turns in the coil. Thus, in order to further reduce the amount of energy that is transmitted to the defibrillation coil **60**, the turns of the defibrillation coil **60** may be tightly wound to maximize the inductance of the coil. Also, a unifilar coil may be used to minimize the space between adjacent turns and maximize the number of turns in the defibrillation coil **60**. In some embodiments, filar of the defibrillation coil **60** has a diameter in the range of about 0.005 to 0.012 inch (about 0.125 mm to 0.305 mm).

In one exemplary implementation, the defibrillation coil **60** has a length of about 80 mm and an outside diameter of about 2.5 mm. The defibrillation coil **60** is a unifilar coil having a filar diameter of about 0.10 mm. The length L of the defibrillation coil **60** that is insulated is about 30 mm, and the distance separating the defibrillation coil **60** from the ring electrode **62** is about 12.5 mm. A lead **14** having this arrangement showed a reduction in heating of about 5-10° C. at the insulation-exposed coil interface of the defibrillation coil **60** compared to leads including a defibrillation coil **60** without a proximal insulated portion.

In alternative embodiments in which the defibrillation coil **60** is multifilar and/or in which the turns of the defibrillation coil **60** are not tightly wound, the length L of the defibrillation coil **60** that is insulated may be increased to increase the impedance of the insulated length L.

Thus, the length L, the number of turns, and the number of filars in the insulated section of the defibrillation coil **60** are selected to provide a reduction in MRI-induced energy in the exposed (i.e., non-insulated) section of the defibrillation coil **60** while minimizing the increase in resistance prior to the exposed portion of the defibrillation coil **60**. In some embodiments, these parameters are selected to provide a total DC resistance in the conductive cable **90** and the insulated length L of less than about 5 Ω.

FIG. **3**A is a cross-sectional view of a distal portion of a lead **100** according to another embodiment of the present invention. FIG. **3**B is a cross-sectional view of a portion of the lead **100** proximal to the distal portion of the lead shown in FIG. **3**A. The lead **100** is another exemplary configuration that may be employed as lead **14** in FIG. **1**. As is shown, the proximal end of the distal portion of lead **100** shown in FIG. **3**A is electrically coupled to the distal end of the proximal portion of lead **100** shown in FIG. **3**B.

The lead **100** includes a distal defibrillation coil **102**, a proximal defibrillation coil **104**, a ring electrode **106**, and a tip electrode **108**. The distal defibrillation coil **102** and proximal defibrillation coil **104** may be used to deliver a high voltage therapy signal to different portions of the heart **18**. The ring electrode **106** and/or the tip electrode **108** may be used for pacing, sensing, or both. In the embodiment shown, the ring electrode **106** is common with the distal defibrillation coil **102** and the tip electrode **108** includes a fixation helix. By making the ring electrode **106** common with the distal defibrillation coil **102**, the diameter of the lead **100** is minimized. When shock therapy is not being delivered through the defibrillation coils **102** and **104**, a pacing voltage may be generated between the electrodes **106** and **108**. In alternative embodiments, the pacing or sensing electrodes are located elsewhere on the lead **100**. The lead **100** may also alternatively include fewer or more electrodes.

The tip electrode **108** is coupled to a conductive coil **110**, which is surrounded by an insulative layer **112** to insulate the conductive coil **110** from other elements of the lead **100**. In some embodiments, the insulative layer **112** extends from the proximal end to the distal end of the lead **100**. The insulative layer **112** may be comprised of, for example, silicone material, Teflon, expanded polytetrafluoroethylene (ePTFE), polytetrafluoroethylene (PTFE), ethylene tetrafluoroethylene (ETFE), or another suitable non-conductive material. The electrodes **106** and **108**, the conductive coil **110**, and the insulative layer **112** combine to form the low voltage pacing/sensing portion **114** of the lead **100**.

The conductive coil **110** extends through the lead **100** and is adapted for connection to the pulse generator **12** at the proximal end of the lead **100**. In some embodiments, the conductive coil **110** is coupled to a proximal connector at the proximal end of the lead **100**. The connectors at the proximal end of the lead **100** are sized and shaped to interface with a connector block or other component of the pulse generator **12**.

The distal defibrillation coil **102** is coupled to a conductive cable **120**, and the proximal defibrillation coil **104** is coupled to a conductive cable **122**. The conductive cables **120** and **122** extend through the lead **100** and are adapted for connection to the pulse generator **12** at the proximal end of the lead **100**. In some embodiments, the conductive cables **120** and **122** may extend through the lead **100** in separate lumens parallel to the conductive coil **110**. The conductive cable **120** is surrounded by an insulating layer **112** at an exterior surface of the lead **100**. In some embodiments, the conductive cables **120** and **122** are each coupled to a proximal connector at the proximal end of the lead **100** that is sized and shaped to interface with a connector block or other component of the pulse generator **12**. The conductive cables **120** and **122** delivers a high voltage defibrillation signal from the pulse generator **12** to the defibrillation coils **102** and **104**, respectively.

To reduce the RF current that is transmitted to the defibrillation coils **102**, **104** by the conductive cables **120** and **122**, a length L_1 of the defibrillation coil **102** and a length L_2 of the defibrillation coil **104** are insulated at the exterior surface of the lead **100** by insulating layer **124**. The insulated lengths L_1, L_2 act as RF filters between the conductive cables **120**, **122** and the exposed portions of the defibrillation coils **102**, **104**, respectively. Consequently, the insulated lengths L_1, L_2 reduce the amount of MRI-induced energy that is transmitted to the defibrillation coils **102**, **104**, respectively. In addition, the portion of the defibrillation coil **102** that is insulated between the ring electrode **106** and the exposed portion of the defibrillation coil **102** may provide a further reduction in the amount of MRI-induced energy that is transmitted to the defibrillation coil **102** and/or the ring electrode **106**.

The inductance of a coil is also directly proportional to the square of the number of turns in the coil. Thus, in order to further reduce the amount of energy that is transmitted to the defibrillation coils **102**, **104**, the turns of the defibrillation coils **102**, **104** may be tightly wound to maximize the inductance of the coil. Also, unifilar coils may be used to minimize the space between adjacent turns and maximize the number of turns in the defibrillation coils **102**, **104**. In alternative embodiments in which the defibrillation coil is multifilar and/or in which the turns of the defibrillation coil are not tightly wound, the lengths L_1, L_2 of the defibrillation coils **102**, **104** that are insulated may be increased to increase the impedance of the insulated lengths L_1, L_2.

Thus, the lengths L_1, L_2, the number of turns, and the number of filars in the insulated section of the defibrillation coils **102**, **104** are selected to provide a reduction in MRI-induced energy in the exposed (i.e., non-insulated) sections of the defibrillation coils while minimizing the increase in resistance prior to the exposed portion of the defibrillation coils **102**, **104**. In some embodiments, these parameters are selected to provide a total DC resistance of the conductive cable **120** and the insulated length L_1 of less than about 5 Ω, and a total DC resistance of the conductive cable **122** and the insulated length L_2 of less than about 5 Ω.

In an alternative embodiment, the diameters of the exposed portions of the defibrillation coils **102**, **104** may be increased to make the outer diameter of the defibrillation coils **102**, **104** substantially equal to the outer diameter of the insulating layer **124**. As a result, the lead **100** has a uniform outer diameter along its length. In one example implementation, a second, larger coil having an outer diameter substantially equal to the outer diameter of the insulating layer **124** is

arranged around and in contact with the exposed portion of each of the defibrillation coils **102**, **104**. In another example implementation, the thickness of the insulating layer **112** is increased under the exposed portion of each of the defibrillation coils **102**, **104** to increase the outer diameter of the defibrillation coils **102**, **104** to be substantially equal to the outer diameter of the insulating layer **124**.

FIG. **4** is a schematic view of an implantable medical device **150** including a pulse generator **12** and a lead **152** according to a further embodiment of the present invention. The lead **152** includes features that may be incorporated into either lead **14** or lead **100** described above. The lead **152** includes a defibrillation coil **154**, a ring electrode **156** and a tip electrode **158**. The defibrillation coil **154**, the ring electrode **156**, and the tip electrode **158** are located near a distal end portion of the lead **152**. The lead **152** may also alternatively include fewer or more electrodes.

The defibrillation coil **154** is coupled to a conductive cable **160**, which extends through the lead **150** and is adapted for connection to the pulse generator **12** at the proximal end of the lead **150**. While not shown in FIG. **4**, electrodes **156** and **158** are also electrically coupled to the pulse generator, such as via conductive coils as shown in FIGS. **2**, **3**A, and **3**B, for example. In some embodiments, the conductive cable **160** is coupled to a proximal connector at the proximal end of the lead **14** that is sized and shaped to interface with a connector block **162** or other component of the pulse generator **12**. The conductive cable **160** delivers a high voltage defibrillation signal from the pulse generator **12** to the defibrillation coil **154**. The lead **152** is arranged in the heart **18** such that the signal delivered by the defibrillation coil **154** depolarizes a critical mass of the heart muscle, terminates an arrhythmia, and allows normal sinus rhythm to be reestablished.

As in the embodiments described above, a length L of the defibrillation coil **154** is insulated to increase the inductance between the conductive cable **160** and the exposed portion of the defibrillation coil **154**. In the embodiment shown in FIG. **4**, high impedance coils **170** are coupled in series with the conductive cable **160**. The high impedance coils **170** provide an additional reduction in the amount of MRI-induced energy that is transmitted to the defibrillation coil **154**. In addition, a high impedance coil **170** may be connected in series with the conductive cable **160** proximate the pulse generator **12** to reduce heating of the pulse generator housing. The high impedance coils **170** may be arranged periodically along the length of the conductive cable **160**. In some embodiments, the length of the conductive cable **160** between adjacent high impedance coils **170** is less than one quarter wavelength ($\lambda/4$) of a signal carried by the conductive cable **160**. This minimizes the energy picked up by the conductive cable **160** in an MRI environment.

In summary, embodiments of the present invention relate to a medical device lead including a proximal connector configured to couple the lead to a pulse generator, an insulative lead body extending distally from the proximal connector, a conductor assembly extending distally from the proximal connector within the lead body. The conductor assembly includes a conductor having a proximal end electrically coupled to the connector and a distal end electrically coupled to a defibrillation coil. A first portion of the defibrillation coil is exposed at an outer surface of the medical device lead and a second portion of the defibrillation coil is insulated at the outer surface of the medical device lead. The insulated portion of the defibrillation coil, which has a high impedance due to its relatively large diameter, acts as a filter for the radio frequency (RF) energy that is picked up by the conductor in a magnetic resonance imaging (MRI) environment. This reduces the transfer of RF energy to the defibrillation electrode, thereby decreasing the amount of heating of the tissue around the electrode.

Various modifications and additions can be made to the exemplary embodiments discussed without departing from the scope of the present invention. For example, while the embodiments described above refer to particular features, the scope of this invention also includes embodiments having different combinations of features and embodiments that do not include all of the described features. Accordingly, the scope of the present invention is intended to embrace all such alternatives, modifications, and variations as fall within the scope of the claims, together with all equivalents thereof.

We claim:

1. A medical device lead comprising:

a first connector configured to couple the lead to a pulse generator;

an insulative lead body extending distally from the first connector;

a first cable conductor extending distally from the first connector within the lead body and having a first segment having a proximal end coupled to the first connector, a second segment, and a third segment, the first cable conductor extending in a straight configuration within the lead body along each of the first, second, and third segments;

a first defibrillation coil exposed at an outer surface of the medical device lead, the first defibrillation coil having a proximal end that is electrically connected to a distal end of the third segment of the first cable conductor; and

a first high impedance coil connected in series to a second high impedance coil, a distal end of the first segment of the first cable conductor coupled to a proximal end of the first high impedance coil, a proximal end of the second segment of the first cable conductor coupled to a distal end of the first high impedance coil, a distal end of the second segment of the first cable conductor coupled to a proximal end of the second high impedance coil, and a distal end of the second high impedance coil coupled to a proximal end of the third segment of the first cable conductor, wherein each of the first segment of the first cable conductor, the first high impedance coil, the second segment of the first cable conductor, the second high impedance coil, the third segment of the first cable conductor, and the first defibrillation coil are respectively electrically connected to one another in series, wherein each of the first, second, and third segments of the first cable conductor has a respective length that is less than one quarter of a wavelength of a signal carried by the first cable conductor and each of the first high impedance coil and the second high impedance coil is insulated at the outer surface of the medical device lead and has an impedance greater than the first cable conductor.

2. The medical device lead of claim **1**, wherein the first defibrillation coil is unifilar.

3. The medical device lead of claim **2**, wherein opposing ends of the first defibrillation coil are short circuited.

4. The medical device lead of claim **3**, wherein the insulative lead body comprises a layer of insulation that extends over a portion of the first defibrillation coil and terminates proximally of the outer surface to expose the first defibrillation coil distally of the layer of insulation.

5. The medical device lead of claim **4**, wherein the first high impedance coil and the first cable conductor have a total DC resistance of less than about 5 Ω.

6. The medical device lead of claim **5**, and further comprising:

a second connector configured to couple the lead to the pulse generator;

a second cable conductor having a proximal end electrically coupled to the second connector;

a second defibrillation coil exposed at an outer surface of the medical device lead; and

a third high impedance coil connected between the second conductor and the second defibrillation coil, wherein the third high impedance coil is insulated at the outer surface of the medical device lead and has an impedance greater than the second conductor.

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